International Journal of Mechanical and Production Engineering Research and Development (IJMPERD)
ISSN (P): 2249–6890; ISSN (E): 2249–8001
Vol. 9, Issue 5, Oct 2019, 571–580

THE EFFECT OF INLET FLOW PROFILE, CAROTID BULB DIAMETER AND NON NEWTONIAN BLOOD VISCOSITY ON THE WALL SHEAR STRESS IN A CAROTID ARTERY BIFURCATION MODEL FOR TRANSIENT FLOW

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ABSTRACT

The carotid artery begins at the aorta within the chest and courses up through the neck to the head. The most common site of atherosclerotic plaque development is at the carotid bifurcation. The size, shape and location of the carotid bulbs are unique to each person. The individual geometry of each person's carotid artery has different flow features and the possibility of atherogenesis. In this work, the effect of inlet flow profile and variations in the carotid bulb sizes and the non-Newtonian viscosity behavior of blood on Wall Shear Stress have been studied by using Ansys Fluent 15. The simulation studies have shown that Wall Shear Stress is reduced with the increase in carotid bulb diameter. Also, the value of Wall Shear Stress decreases remarkably when the nature of the inlet velocity profile changes. What is observed is that when the blood is considered Newtonian (in terms of viscosity) and if the velocity profile is parabolic, there are more chances of atherosclerotic plaque formation there.

KEYWORDS: Inlet Flow Profile, Carotid Bulb Diameter, Non-Newtonian Blood Viscosity, Wall Shear Stress & Atherosclerotic Plaque

Received: Jun 09, 2019; **Accepted:** Jun 29, 2019; **Published:** Sep 20, 2019; **Paper Id.:** IJMPERDOCT201950

Original Article

NOMENCLATURE

CCA-Common Carotid Artery

ICA-Internal Carotid Artery

ECA-External Carotid Artery

UDF-User Defined Function

WSS-Wall Shear Stress

1. INTRODUCTION

The carotid artery begins at the aorta within the chest as CCA and courses up through the neck to the head. Near the larynx, the CCA divides into the ECA and ICA.

The ECA supply blood to the face. The ICA supply blood to the brain as shown in figure 1 [1]. Flow in bifurcating channels involves various issues to be inscribed, whether the fluid flowing is Newtonian or non-Newtonian, channel wall is rigid or elastic, fluid properties, channel dimension and cross-section.

The above-mentioned parameters have a direct effect on velocity profile, wall shear stress (WSS) distribution and pressure distribution. While dealing with biological applications such as blood flow in the artery the above-mentioned parameters have a direct relation to the health of the artery.

Moreover, to achieve detailed information on velocity, WSS and pressure distribution, further research needs to be carried out. The most habitual site of atherosclerotic plaque buildup is at the carotid bifurcation. Mahesh and Gupta have studied that, by numerically modelling the flow behavior in the carotid sinus over a range of possible locations, sizes, and shapes [1]. The effect of bifurcation angle, off plane angle and the effect of symmetric bifurcations on WSS have been studied [2]. To the best of authors' knowledge, there is no study available in the literature which reported the effect of the sinus size, inlet velocity profile and non-Newtonian on the flow behavior and wall shear stress. The present work addresses this gap by numerically modeling the effect of inlet flow profile and variations in the carotid bulb sizes and the non-Newtonian viscosity behavior of blood on Wall Shear Stress.

2. LITERATURE OVERVIEW

Although smoking and hypertension are some of the main causes of atherosclerosis, it generally happens when wall shear stress is less than 0.4 Pa [3]. Malek et al. have reported that the outer walls of the vessel bifurcation are distinguished by low wall shear stress and are more prone to the plaque formation [3]. Ku concluded that blood flow in arteries is dominated by unsteady flow phenomena [4]. The usual arterial flow is laminar with peripheral flows produced at curves and branches and velocity profile bending can create pockets, in which the regulation of the wall shear stress swings. The data obtained from hemodynamics in pulsatile flow tells that there are the geometric features that will cause the variation in shear stress located on a particular site on the wall vessel from one individual to another. Such shear stress is responsible for the variation in the location and rate of development of plaque formation [5]. Rindt et al. [6] have measured the axial velocities in an expanded, two-dimensional, rigid model of the carotid artery bifurcation model using an experimental method for both steady and unsteady flow conditions. They had also developed a numerical model. Their study concluded that the bifurcation had a strong effect on the upstream flow in CCA and high-velocity gradients are observed at ICA and CCA.

Schulz et al. [7] conclude that the important thing of geometric liableness for atherosclerosis is the presumption that vessel geometry alters adequately widely across the population.

In the modern study of angiogram reports from nearly 3000 patients says that there was a crucial imbalance in the ratios of diameter and area in the carotid bifurcation from person to person. Perktold et al. [8, 9] had developed a carotid artery model and compared the velocity and wall shear stress value with their experimental readings with good agreement. Younis et al. [10] conclude that the lower wall shear stresses occur at the outer wall of carotid bulbs and higher wall shear stress occurs at the external carotid artery. Gijsen et al. [11] had conducted Laser Doppler anemometry experiments of steady flow for carotid artery bifurcation to investigate the influence of non-Newtonian properties of blood on the velocity distribution and the axial velocity field of the non-Newtonian fluid was flattened, had lower velocity gradients at the divider wall and higher velocity gradients at the non-divider wall. Comparison between numerical and experimental methods had shown a good agreement. Steinman et al. [12] had found a relationship between local hemodynamics and plaque progression as it was difficult to monitor these factors in humans. Recently Sahoo et al. [13] assessed the hemodynamics of carotid artery models with carotid bulbs of various sizes, the oscillatory shear index (OSI) and flow velocity distribution were evaluated in carotid models without a carotid bulb and with carotid bulbs of known geometry and concluded that the relation between the OSI and the carotid bulb size could serve as a risk indicator for atherosclerosis.

3. METHODOLOGY

A. Geometry Creation

The carotid artery is assumed to be rigid and having the same radius throughout in each except for the sinus in ICA [2]. The diameter of CCA is taken to be 8 mm and length 41 mm. The ECA has a diameter of 4.62 mm and length 47 mm, and ICA has a diameter of 5.55 mm and a length of 47 mm. The ICA sinus is represented by an ellipsoid having a major axis of 22 mm. The sinus has the same axis as that of ICA. The bifurcation has been considered to be symmetric and the bifurcation angle (α) i. e. the angle between ICA and ECA has been assumed to be $60^{o.}$

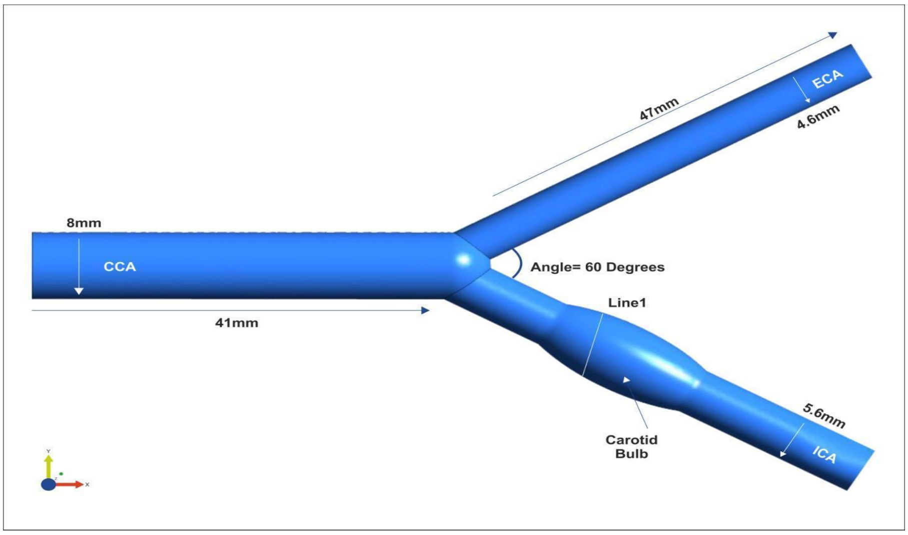


Figure 1: Carotid Artery Bifurcation.

B. Mathematical Model

1. Governing Equation and Boundary Condition

The blood flow in the carotid artery is laminar, For the unsteady laminar flow of an incompressible, Newtonian fluid, the mass and momentum conservation equations can be written as the following Equations (1–2) [14].

Continuity equation:

$$\nabla . v = 0 \quad (1)$$

Momentum conservation equation:

$$\rho\left(\frac{\partial v}{\partial t} + \nabla . vv\right) = -\nabla P + \mu \nabla^2 v \ \rho\left(\frac{\partial v}{\partial t} + \nabla . vv\right) = -\nabla P + \mu \nabla^2 v \quad (2)$$

At the inlet boundary, the flow was specified to be time-dependent and periodic such that it mimics blood flow in the carotid artery. Blood flow at the inlet is considered as pulsatile and parabolic velocity profile by plugging UDF. At the outlet boundary, the pressure is specified to be zero (gauge). No slip boundary condition has been specified on all the walls.

A coupled scheme is used for the pressure-velocity coupling and momentum equation has been discredited using the second-order upwind scheme. The time step has been chosen to be 0.0545 s, i. e. 300 iterations per cycle. The density and viscosity of blood are 1060 kg/m^3 and 0.0035 Pa.s, respectively [8].

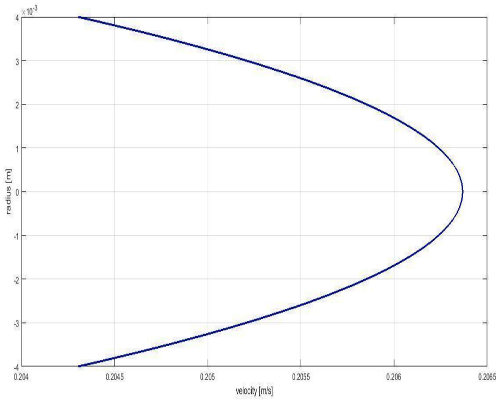


Figure 2: Parabolic Velocity Profile.

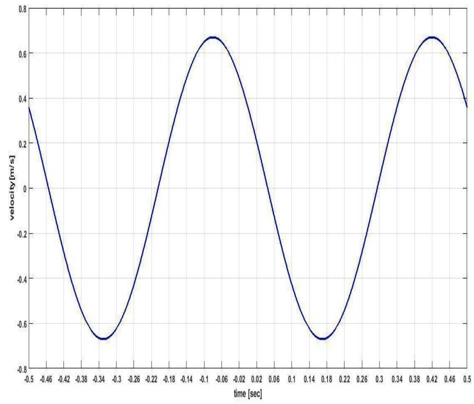


Figure 3: Pulsatile Velocity Profile.

4. RESULTS AND DISCUSSIONS

Three different geometries of carotid bulb diameters of 8, 10, 12 mm are taken with carotid bulb located on Internal Carotid Artery at length of 20mm is considered. The effect of the above factors on wall shear stress, velocity profiles and on the vorticity are investigated. Simulations have run up to 0.22 seconds, 4 cycles for each case. Velocity profiles are calculated at line 1 for different carotid bulb diameters, for Newtonian and Non-Newtonian (Carreau type) behaviors of pulsatile and parabolic velocity inlet profiles.

A. Effect on the Velocity Distribution

Figures 4–6 show that velocity vs distance of different carotid bulb diameters drawn at line 1. As the shear stress is directly proportional to velocity gradient, as the bulb diameter increases the velocity gradient decreases and thereby resulting in the decrease of wall shear stress. When the Wall Shear Stress is less than 0.4 Pa, there will be more chance of atherosclerosis formation [3]. When the carotid bulb diameter is 8 mm, its Wall shear stress value is greater than 0.4 Pa for Pulsatile flow of both Newtonian and Non-Newtonian nature of the blood flow. So there will be less risk of plaque formation when the bulb diameter is less. When the bulb diameter increases, the slope of the pulsatile nature increases.

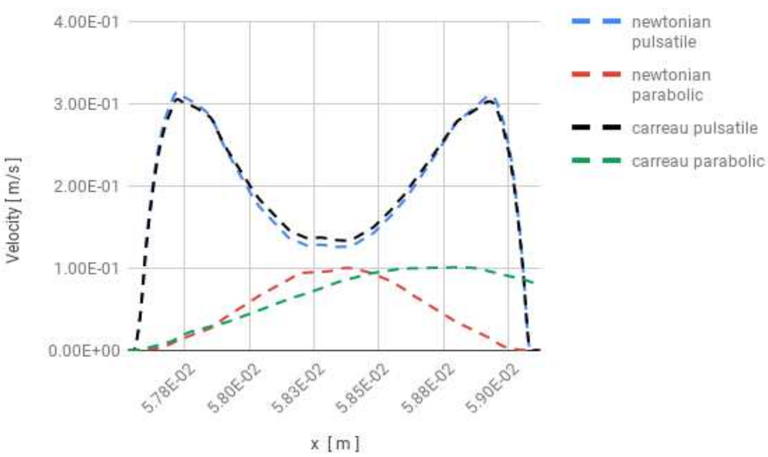


Figure 4: Velocity Vs Distance Drawn at Line 1 for 8 mm.

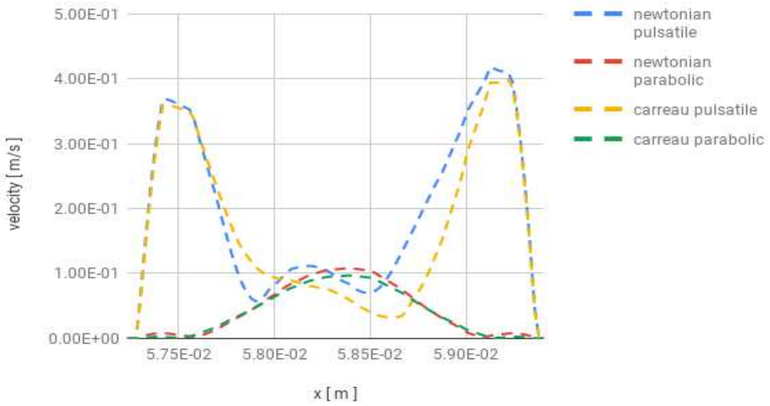


Figure 5: Velocity Vs Distance Drawn at Line 1 for 10 mm.

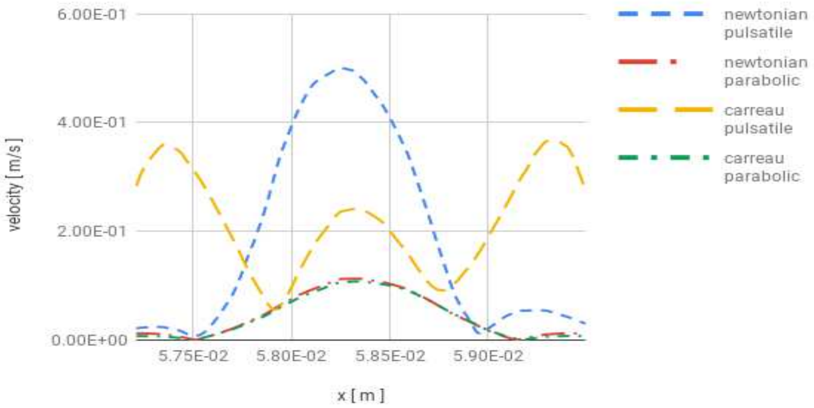


Figure 6: Velocity Vs Distance Drawn at Line 1 for 12 mm

B. Effect on Wall Shear Stress

As the bulb diameter increases the wall shear stress value decreases in every case whether it is Newtonian or non-Newtonian nature and parabolic or pulsatile flow type. For any carotid bulb diameter, the lowest wall shear stress will encounter for the Newtonian parabolic profile type. The lowest wall shear stress is encountered for the bulb diameter of 12 mm for the Newtonian parabola profile type. Figures 7–12 show how the wall shear stress values vary with changes in bulb diameter.

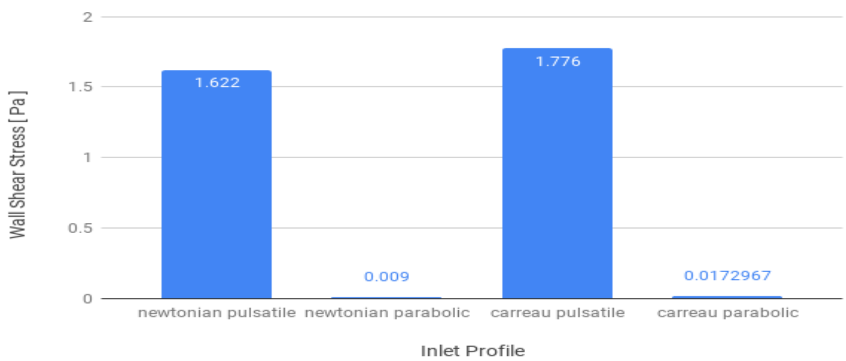


Figure 7: 8 mm Wall Shear Stress Distribution.

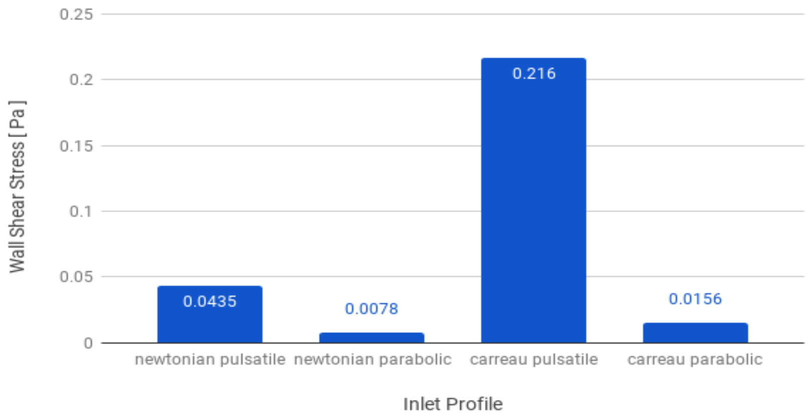


Figure 8: 10 mm Wall Shear Stress Distribution.

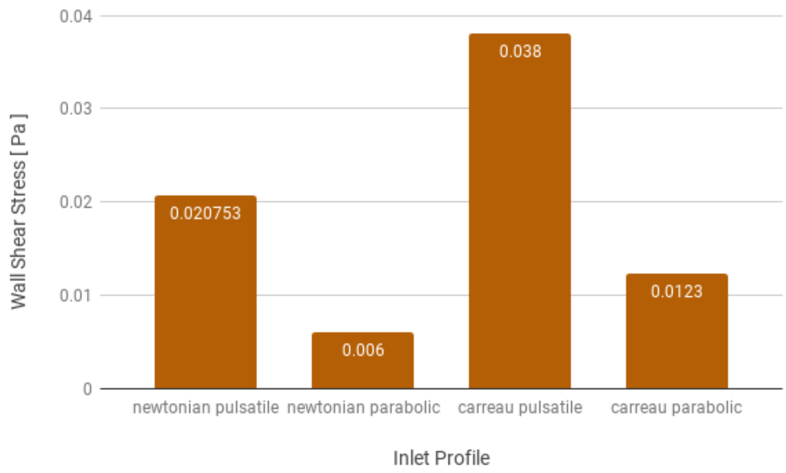


Figure 9: 12 mm Wall Shear Stress Distribution.

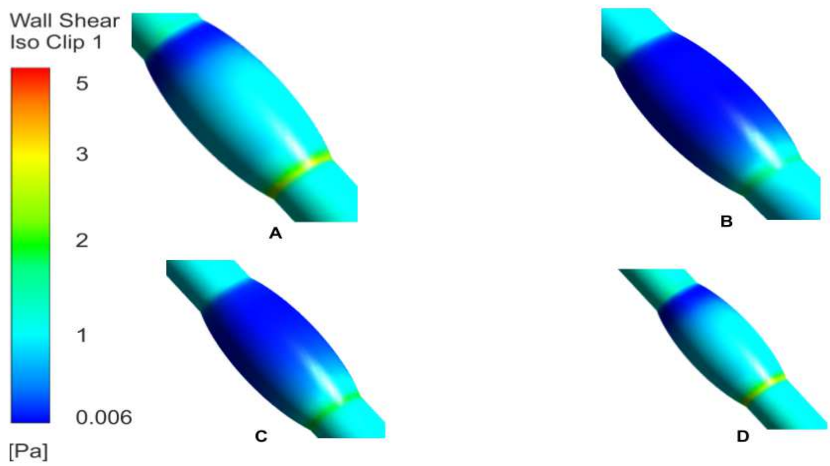


Figure 10: Wall Shear Stress Contours of 8 mm (A) Carreau Pulsatile Model (B) Newtonian Parabolic Model (C) Carreau Parabolic Model (D) Newtonian Pulsatile Model.

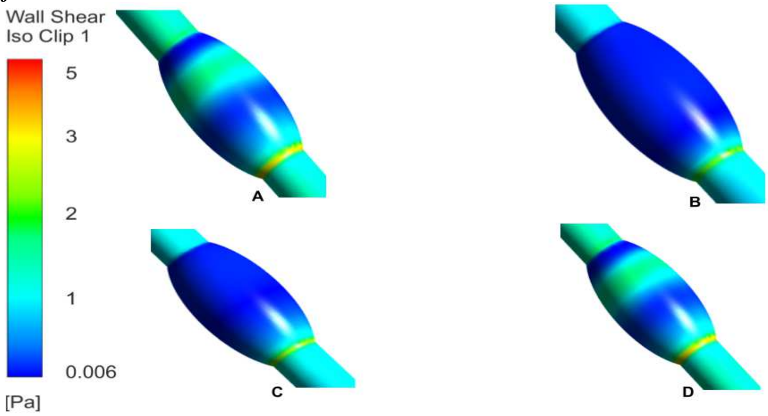


Figure 11: Wall Shear Stress Contours of 10 mm (A) Carreau Pulsatile Model (B) Newtonian Parabolic Model (C) Carreau Parabolic Model (D) Newtonian Pulsatile Model

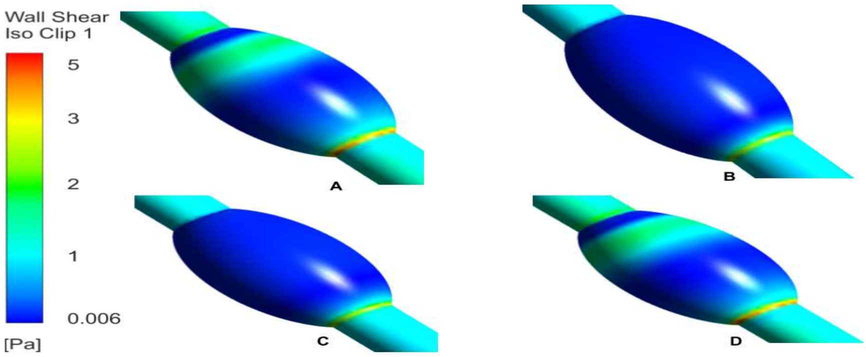
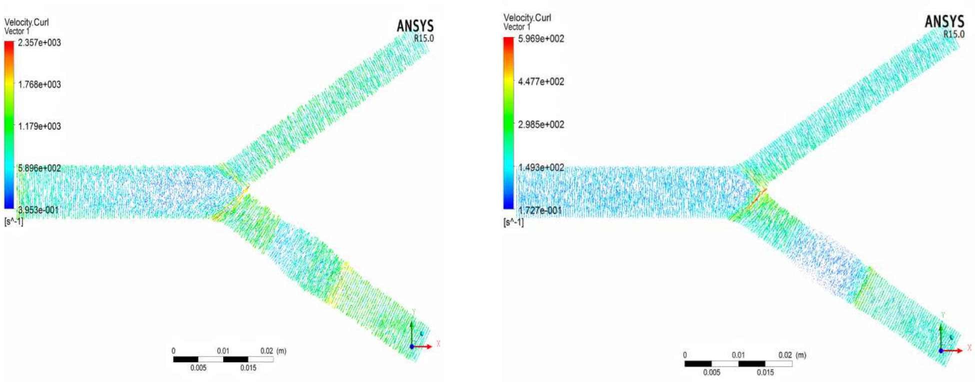


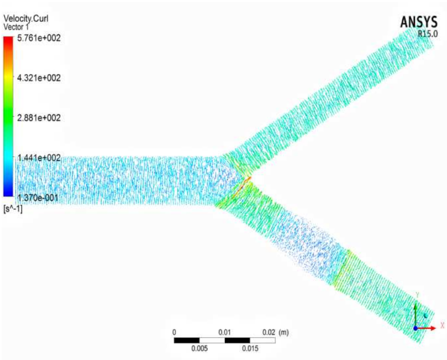
Figure 12: Wall Shear Stress Contours of 12 mm (A) Carreau Pulsatile Model (B) Newtonian Parabolic Model(C) Carreau Parabolic Model (D) Newtonian Pulsatile Model

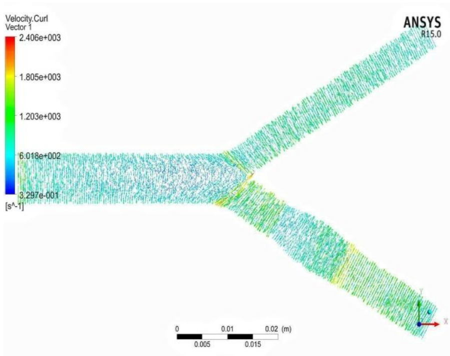
C. Effect on Vorticity

The patterns of the vorticity contours have been shown to have back-flow both upstream and downstream and several flow separation zones at the various critical zones. It was observed that the increase in the carotid bulb diameter increases the recirculation area and backflow as shown in figures 13–15. For the same flow rate, the velocity magnitude decreases with an increase in the carotid bulb diameter. Consequently, the wall shear stress decreases with an increase in the carotid bulb diameter.



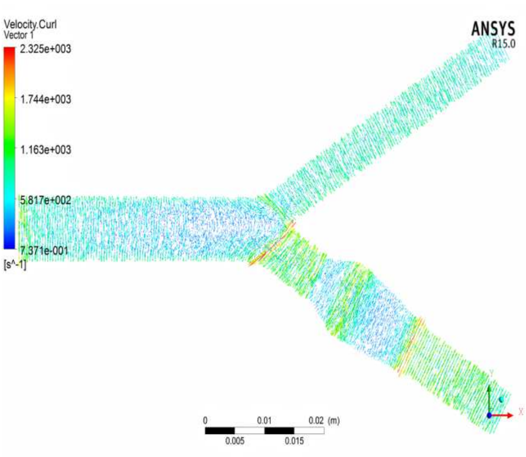
(A) Carreau Pulsatile. **(B) Newtonian Parabolic Model.**

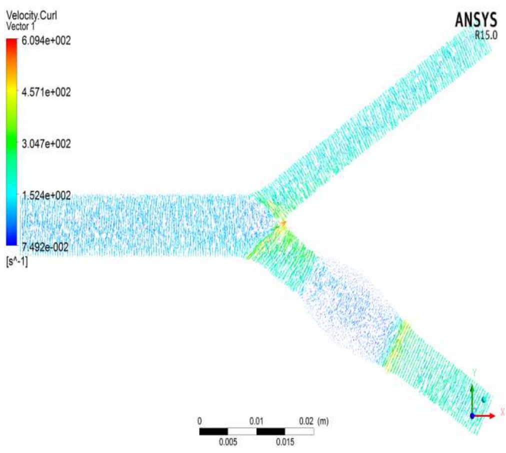




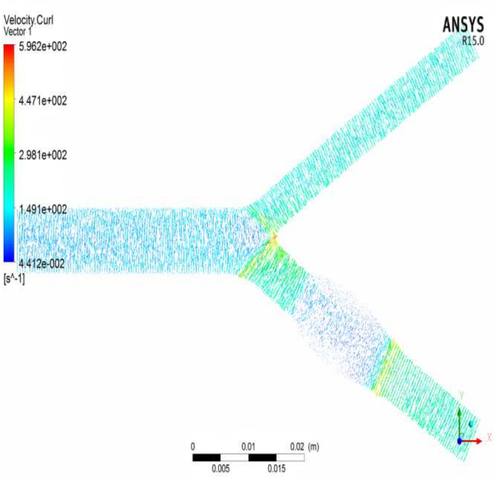
(C) Carreau Parabolic Model. **(D) Newtonian Pulsatile Model.**

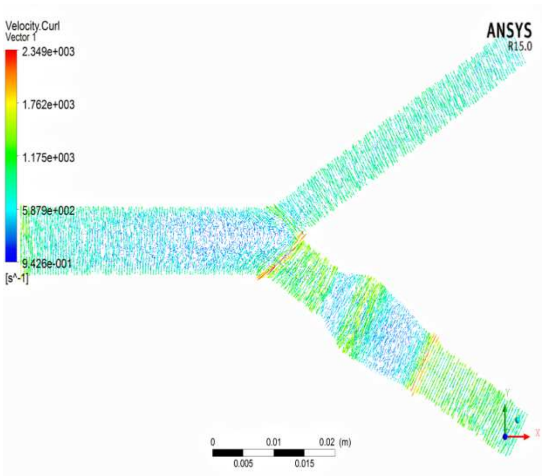
Figure 13: Vorticity Vectors of 8 mm Model.





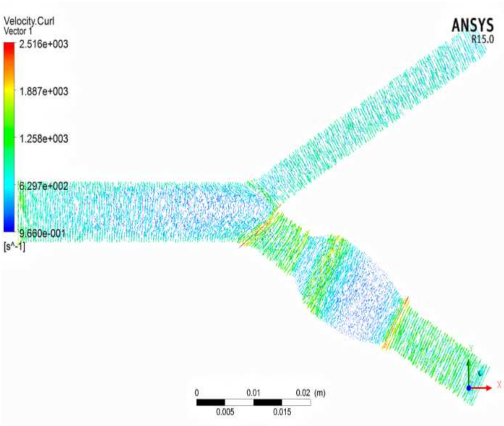
(A) Carreau Pulsatile Model **(B) Newtonian Parabolic Model**



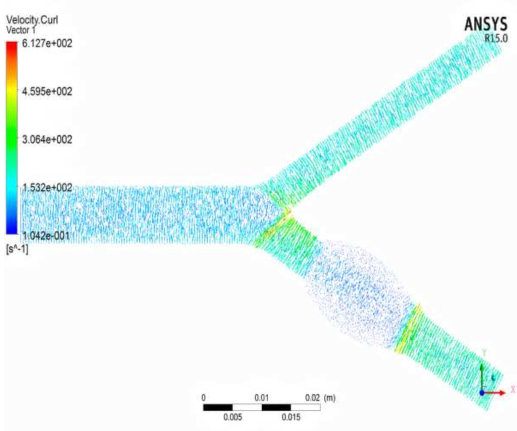


(C) Carreau Parabolic Model **(D) Newtonian Pulsatile Model**

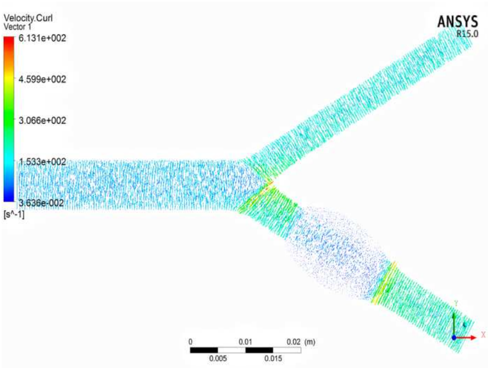
Figure 14: Vorticity Vectors of 10 mm



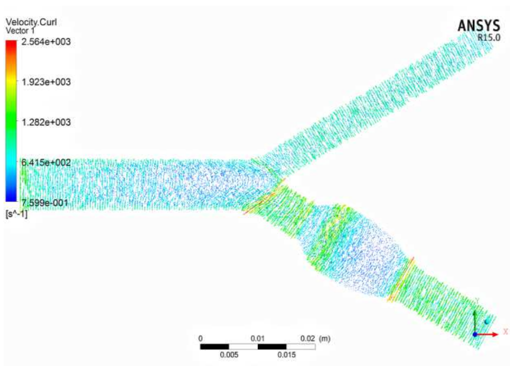
(A) Carreau Pulsatile Model.



(B) Newtonian Parabolic Model.



(C) Carreau Parabolic Model.



(D) Newtonian Pulsatile Model.

Figure 15: Vorticity Vectors of 12 mm.

5. CONCLUSIONS

From the exhaustive investigations, it is found that the increase in carotid bulb diameter increases the recirculation and thereby decreases the wall shear stress. The main aim of the present study was on the flow in the carotid artery, where atherosclerotic plaque builds up and on the effects of the Newtonian and non- Newtonian nature and parabolic and pulsatile flow behavior. Nevertheless, the risk of atherogenesis is related to the flow in the external carotid artery and at the bifurcation area may be important. Also, the location of the carotid bulb diameter, which is fixed in this study, may significantly affect the flow patterns and the risk of the disease. The work done in this research work, however, can be used to quantify the risk from low WSS irrespective of the location and shape of the carotid bulb diameter. It is also suggested to consider the location of the carotid bulb diameter as a variable and carry this work forward.

REFERENCES

1. *Mahesh Nagargoje and Raghavendra Gupta, 2018. Effect of Sinus Shape and Position on Pulsatile blood flow in Carotid artery bifurcation, Proceedings of the 7th International and 45th National Conference on Fluid Mechanics and Fluid Power (FMFP) December 10–12, 2018, IIT Bombay, Mumbai, India.*

2. *Khan, S. A., Shahzer, M., Sultan, D., & Ali, R. Thermal And Solutal Buoyancy Effects On Mixing Of Opposed Laminar Jets In A Two-Dimensional Passive Mixer At A Higher Reynolds Number.*

3. K. T. Nguyen, C. D. Clark, T. J. Chancellor and D. V. Papavassiliou; 2008. Carotid geometry effects on blood flow and on risk for vascular disease, Journal of Biomechanics 41, 11–19.

4. A. M. Malek, S. L. Alper and S. Izumo, 1999. Hemodynamic shear stress and its role in atherosclerosis. Journal of the American Medical Association 282, 2035–2042.

5. Alex, R. A. (2014). Personality pattern of female cardiovascular patients and cancer patients: An analytical study. Int J Res Appl Nat Soc Sci, 2(3), 61–72.

6. Ku, D. N., 1997. Blood flow in arteries. Annual Review of Biomedical Engineering 29, 399–434.

7. Friedman, M. H., Deters, O. J, Mark, F. F. and Bargeron, C. B., Hutchins GM. Arterial geometry affects hemodynamics. A potential risk factor for atherosclerosis. Atherosclerosis. 1983; 46:225–231.

8. C. C. M. Rindt, A. A. V. Steenhoven and R. S. Reneman; 1987. A numerical and experimental analysis of the flow field in a two-dimensional model of the human carotid artery bifurcation. J. Biomechanics 20, 499–509.

9. Roy, S. S. Monstrosity and the multi-dimensional other in mary shelley's frankenstein.

10. Schulz, U. G. and Rothwell, P. M., Major variation in carotid bifurcation anatomy: a possible risk factor for plaque development? Stroke. 2001; 32:2522–2529.

11. K. Perktold, R. O. Peter, M. Resch, G. Langs; 1991. Pulsatile non-Newtonian blood flow in three-dimensional carotid artery bifurcation models: a numerical study of flow phenomena under different bifurcation angle, J. Biomed. Eng, vol. 13.

12. Areo, M. O. Regbe-Regbe: Multidimensional Impact of Cloth and Colour in Ojude-Oba Festival.

13. Perktold, K., Hilbert, D., 1986. Numerical simulation of pulsatile flow in a carotid bifurcation model. Journal of Biomedical Engineering 8, 193–199

14. Younis, H. F., Kaazempur-Mofrad, M. R., Chan, R. C., Isasi, A. G., Hinton, D. P., Chau, A. H., Kim, L. A., Kamm, R.D., 2004. Hemodynamics and wall mechanics in human carotid bifurcation and its consequences for atherogenesis: investigation of inter-individual variation. Biomechanics and Modeling in Mechanobiology 3 (1), 17–32.

15. Gijsen, F. J. H., van de Vosse, F. N., Janssen, J. D., 1999. The influence of the non-Newtonian properties of blood on the flow in large arteries: steady flow in a carotid bifurcation model. Journal of Biomechanics 32 (6), 601–608.

16. Steinman, D. A., Thomas, J. B., Ladak, H.M., Milner, J.S., Rutt, B.K., Spence, J.D., 2002. Reconstruction of carotid bifurcation hemodynamics and wall thickness using computational fluid dynamics and MRI. Magnetic Resonance in Medicine 47 (1), 149–159.

17. T. Saho, H. Onishi, 2017. Quantitative analysis of hemodynamic stress on temporal variations of cardiac phases in models of human carotid bulbs. Radiol Phys Technol 10, 475–482.

18. Frank M. White, Fluid Mechanics, McGraw Hill, 2011.